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Fatigue-Related Changes in the Coordination of Lifting and Their Effect on Low Back Load

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ABSTRACT. In this study, changes in movement coordination caused by fatigue that developed during repetitive lifting were examined. Five men performed 6 times a 5-min bout of lifting an 8-kg barbell at 15 lifts/min, using two lifting techniques; one minimized trunk rotation (squat lift), and the other minimized rotation in the knee joint (stoop lift). Kinematics and dynamics were studied by means of movement analysis and inverse dynamics, using a two-dimensional linked segment model. Within-subject variation over repetitive lifts of the time course of joint angles was smaller than between-subjects variation on the first analyzed lift. Relative timing between joint rotations did not change significantly across repetitive lifts, except between knee and hip in the squat lift. No change of the lumbosacral torque over repetitive lifts was found. The adaptability of the neural control appeared to be sufficient to accommodate the strong changes of the input–output characteristics of the muscles caused by fatigue so that an essentially constant performance of the movement act was maintained.

Key words: biomechanics, motor control, muscle fatigue

The act of lifting objects is certainly a physically exerting task, but also, from the point of view of motor control, it is by no means trivial. There are demands for setting the trajectory of the object to be lifted at the same time that balance requirements have to be met. Considerable linear and angular accelerations of body segments must be produced and reversed in time to end the movement while the inertial properties of the lifter–load system are changing as the object is picked up or released. Moreover, the load on the joints and surrounding muscles, especially in the low back, may reach levels that exceed tissue tolerance. Therefore, to avoid damage, one must control the act so that the load on the musculoskeletal system is kept within bounds.

Repetitive lifting has been shown to be a risk factor for the development of low back pain (e.g., Frymoyer et al., 1983). It has been hypothesized that repetitive lifting leads to muscular fatigue and, hence, reduced motor performance and hampered motor control. This, in turn, is expected to lead to increased loads on passive structures in the low back and, consequently, to mechanical damage and, ultimately, low back pain (Jørgensen, Jensen, & Kato, 1990; Parnianpour, Nordin, Kahanovitz, & Frankel, 1988; Roy, DeLuca, & Casavant, 1989; Seidel, Beyer, & Bräuer, 1987). To our knowledge, however, there has been little evidence that supports this hypothesis, because no comprehensive description of fatigue-related changes, if any occur, in the way the lifting act is performed has been provided in the literature.

As a motor control problem, the occurrence of fatigue during repetitive lifting is especially interesting. Scholz (1993a, 1993b) has shown that during a limited number of repeated lifts the kinematics are fairly stable. This suggests that a stereotyped stimulation pattern controls the output of the effector organs, that is, the muscles. After a high number of repetitions, considerable muscle fatigue has been shown to occur (Petrofsky & Lind, 1978; Potvin & Norman, 1993). If stereotyped kinematics are to be maintained, therefore, the changed input–output characteristics of the

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effector organs should be accommodated by adaptation of the activation pattern generated by the nervous system. If these adaptations are insufficient, several changes could take place.

The studies on fatigue development during lifting, mentioned above, have mainly focused on the paravertebral muscles. During lifting, these muscles produce forces that have been estimated to range up to 100% of their maximum, taking into account force length and force contraction velocity relationships (McGill & Norman, 1986). Potvin and Norman (1993) found isometric trunk extensor strength to be decreased by about 20% after repetitive lifting. Keeping in mind that isometric strength underestimates the consequences of fatigue for dynamic force production (Beelen & Sargeant 1991), fatigue could be expected to severely influence trunk dynamics during lifting. An increase of the mass of the object lifted has been shown to increase the phase lag between the rotation of the upper body and the rotation in the joints of the lower extremity during squat lifting (Schipplein, Trafimow, Andersson, & Andriacchi, 1990; Scholz, 1993a, 1993b). This might indicate that at a high mass of the object, the strength of the paravertebral muscles is insufficient to accelerate the upper body in the initial posture. As a consequence, the flexion angle in the low back even increases during the first part of the lifting movement with heavier weights. Similar changes may occur when the trunk extensor muscles become fatigued. This would be compatible with the above hypothesis, which assumes a higher strain of passive structures of the low back when fatigue occurs.

If the fine-tuning of the activation of the muscles crossing the lumbosacral joint is impaired, trunk extension will probably be less smoothly executed. Though the torque-producing capacity of the muscles is decreased, the net torque at the lumbosacral joint might even increase when a nonoptimal postural configuration is adopted at the start of the movement or the timing of joint rotations over successive joints is impaired.

Toussaint and coworkers (Toussaint, Baar, Langen, Looze, & Diëen, 1992; Toussaint et al., 1995) have shown that control of the ground reaction force through the biarticular leg muscles plays an important role in the coordination of the lifting act. Though somewhat less than the paravertebral muscles, the hamstring muscles and the gastrocnemius muscles show substantial activity during lifting (Looze, Toussaint, Diëen, & Kemper, 1993). Therefore, considerable fatigue can be expected to occur in these muscles as well. Hence, changes of the torques produced at the joints of the lower extremities could also occur during repeated lifting and as such, hamper the control of the movement of the entire body and load.

We performed the present study to see if fatigue affects the execution of lifting tasks and whether any such changes cause the load on the low back to increase. To accomplish this objective, we required subjects to perform repeated lifts. An inverse dynamic analysis of the lifting movements was made, and the between-trials variation of kinematic and dynamic parameters was analyzed.

Method

Five healthy men performed six times a 5-min bout of lifting an 8-kg barbell of at a rate of 15 times/min, interspersed by a 2-min rest. During these 2 min, measurements were made of spinal shrinkage, which have been reported elsewhere (Dieën, Creemers, Draisma, Toussaint, & Kingma, 1994). All subjects were free of low back pain and signed informed consent prior to participation. They were thoroughly acquainted with the tasks before any measurements were taken. On one day, the subjects lifted with the squat technique, in which the knees are flexed at the initiation of the lift while the trunk remains erect as much as possible. On another day, the subjects lifted with a stoop technique, in which the knees remain extended while the trunk and hips are flexed. The initial position of the barbell was standardized at 0.23 m above the ground. Each subject chose, for each technique separately, the minimal horizontal distance of the load from the body that allowed for a smooth performance of the task. The load was lifted following a vertical trajectory (guided by two vertical wants), and the movement was stopped and subsequently reversed when upright stance, with the barbell just in front of the hips, was reached. The elbows remained extended throughout the lifting and lowering cycle.

Ground reaction forces were measured by means of a forceplate (Kistler 9218 B) and sampled at 60 Hz. An automated optoelectronic system (Vicon, Oxford Metrics) traced the position of retroreflective markers placed on the caput of the fifth metatarsal, the lateral malleolus, the lateral femur epicondyle, the upper margin of the greater trochanter, the spinous process of the first thoracic vertebra, and the estimated projection of the lumbosacral joint on the right side of the body (Looze, Kingma, Bussmann, & Toussaint, 1992). Recordings were made of the 9th complete lifting and lowering cycle during the first bout and of 1 of the last 10 cycles during this and each subsequent 5-min bout. The sampling rate was 60 Hz.

The sagittal plane coordinates of the marker positions were low-pass filtered at a cut-off frequency of 5 Hz (zero-phase lag, second-order Butterworth). We calculated time series of joint angles, angular velocities, jerk (that is, the first derivative of angular acceleration), and phase angles to describe the kinematics. Phase angles were calculated, using the procedure described by Kelso, Saltzman, and Tuller (1986). In short, joint angles and angular velocities were normalized so that their maximum value was set to 1 and their minimum value to –1. Subsequently, a plot was constructed of joint angle on the ordinate against angular velocity on the abscissa. The phase angle is the angle between the vector from the origin to a point on the phase plot and the right-hand ordinate. The low angular excursions in the knee and ankle joint when stoop lifting precluded reliable calculation of the time series of the phase angle of these joints. These were therefore not used in the analysis for this lifting technique. The mean squared jerk at each joint over the entire lifting-lowering cycle was used as an indicator of the smoothness of the movements.

Coordination (i.e., the relative timing of joint rotations) was quantified by means of the relative phases of neighboring joints, that is, the phase difference between ankle and knee, knee and hip, and hip and lumbosacral joint. The phase lag between joints was defined as the mean relative phase determined over the lifting and lowering phases separately.

The load on passive structures in the low back depends on the joint torque around the lumbosacral junction and on the way in which this torque is distributed across muscles and passive structures. Obtaining detailed information on load sharing is not feasible in the context of a fatigue study. However, an indication of the contribution of the passive structures can be obtained from the lumbosacral angle. This angle bears a strong relationship to spinal flexion (Kippers & Parker, 1989) and, as such, to the elongation of the osteoligamentous spine. The contribution of these passive structures to the torque is directly related to their elongation (Adams & Hutton, 1986). Because cocontraction of flexor muscles is weak or absent during lifting (McGill & Norman, 1986), the torque and joint angle combined provide a reasonable indicator of the load on the back. The time series of the lumbosacral joint torque was calculated by inverse dynamic analysis, employing the model described by Looze et al. (1992). In view of the deterministic relationship between kinematics and dynamics and the higher reliability of the kinematic data, no other dynamic variables were included in the analysis.

We used analysis of variance with repeated measures to test for the effects of repetition and lifting technique. In the analysis of variance, missing values (3 out of 70 trials) were estimated by an iterative approach, using the procedure described by Healy and Westmacott (1956). In short, the missing value was initially set at the grand mean; subsequently, we calculated treatment effects and used these to update the value for the particular observation. This procedure was repeated until a zero residue for the observation was reached. This procedure allows one to use incomplete data sets of a subject rather than drop the data for the subject. Because the missing observation takes on exactly the value predicted by the treatment effects as estimated from the other observations (zero residue), it does not affect the estimation of the treatment effects. We subtracted the number of missing values from the total number of observations planned to adjust the degrees of freedom of the total sum of squares. Paired t tests were used for post hoc comparisons. In addition, we used Spearman's coefficient of rank-order correlation to test if a systematic change of the values of selected parameters occurred over the repeated lifting bouts. To create plots of group averages of selected parameters, we normalized the length of the time series of these parameters by means of cubic spline interpolation, setting the length of one complete lifting and lowering cycle to 100%.

Results

A complete lifting and lowering cycle took, on average, 4.0 s (*SD*, 0.3 s). The normal coordination during lifting is shown in Figures 1 to 2. Figure 1 presents plots of the excursion of the joint angles from the initiation of the lifting movement through the whole lifting and lowering cycle. The plots represent averages of the first analyzed trial of all subjects for each technique. The standard deviations indicated by the vertical bars clearly show that interindividual variation was considerable. The difference between the lifting techniques can be seen clearly in the plots of the ankle and knee joint excursions. Figure 2 shows the phase lag between neighboring joints. A negative relative phase indicates that the distal joint led the proximal. As can be seen, a distal to proximal sequence of joint rotations was present during lifting with both techniques, whereas during lowering, the opposite sequence occurred. During lowering, the sequencing was less consistent, as evidenced by the high interindividual variation.

It was evident that the subjects did succeed in keeping up with the predetermined pace of lifting, because the duration of the lifting and lowering movement did not change significantly across trials. Figure 3 presents superimposed plots of the time course of the joint angles of 1 subject, giving a typical example. The angular excursions of the ankle and knee in stoop lifting were not analyzed in this way, in view of the limited movement that occurred at these joints. As can be seen from the plots, the variability across repetitive lifts was limited. At each point of the normalized time axis, the standard deviation across trials was calculated for each subject. The median values thereof are listed in Table 1. For comparison, the results of the same analysis performed across subjects for the first analyzed trial are also given. Clearly, within-subject variation across repetitive lifts was smaller than between-subjects variation for the first analyzed lift. We used analysis of variance to test whether the minimum joint angles, that is, the joint angles near the initiation of the lift, changed across repetitive trials. No significant effect was found. Of course, the differences between the two techniques were highly significant ($p <$ .001 for all four joints).

The group averages of the jerk tended to increase in all joints and both techniques. The analysis of variance indicated, however, that the differences between trials were not significant. Moreover, the increase was not a consistent finding in all subjects. Rank-order correlations on individual data for squat lifting were significant in only one case in the ankle, none in the knee, one case in the hip, and one in the lumbosacral junction. For stoop lifting, only 1 subject showed a significant rank-order correlation between time and jerk in the hip and none in the lumbosacral junction.

The coordination of the lifting act as described by the relative phase of the joint rotations (presented in Figure 4)

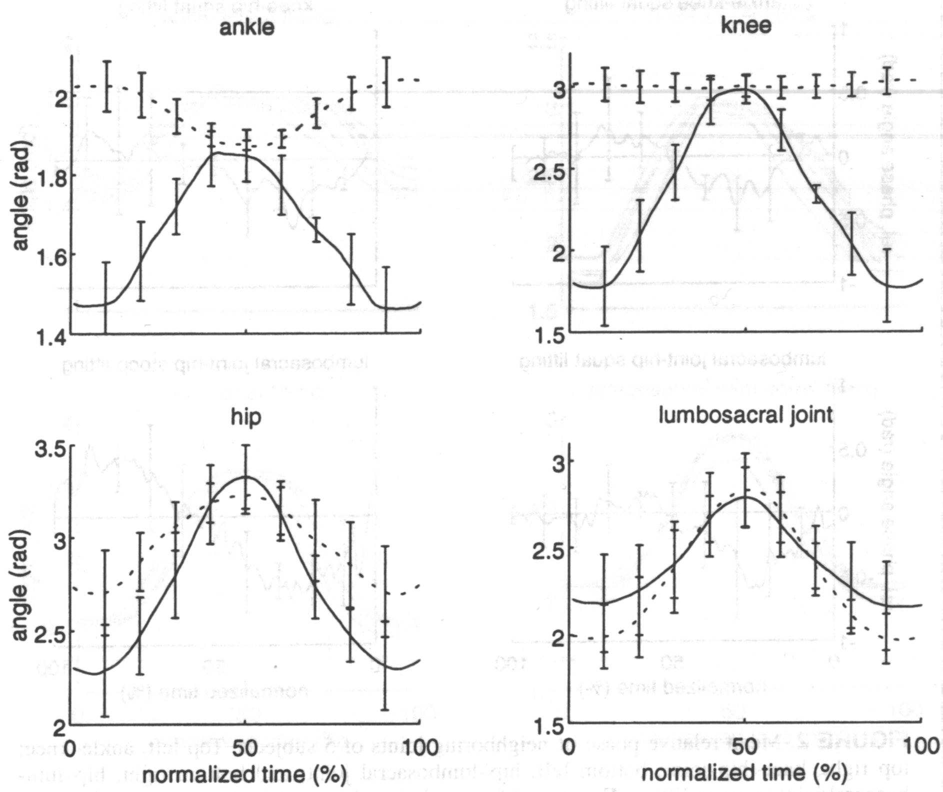

FIGURE 1. Mean joint angle time series of 5 subjects. The solid lines represent the squat lift, and the dashed lines represent the stoop lift; bars indicate plus and minus one standard deviation. Top left, ankle joint; top right, knee joint; bottom left, hip joint; and bottom right, lumbosacral junction.

showed considerable within-subject variation. The median standard deviations within subjects across repetitive lifts and between subjects for the first analyzed lift are listed in Table 2. The mean phase lags for lifting and lowering are presented in Figures 5 and 6, respectively. Analysis of variance revealed a significant effect of trial on the phase lag of the knee and hip during lifting with the squat technique, F (6, 22) = 3.0, $p = .028$. Post hoc testing showed the difference between Trial 1 and all other trials and the difference between Trials 2 and 3, on one hand, and Trial 7, on the other hand, to be significant. Spearman's coefficient of rank correlation indicated the decrease of the phase lag to be systematic, when looking at the group means ($r = .82$). When looking at the individual data, the decrease appeared to be systematic in 2 out of 5 subjects (rs = .79 and .90). In the other subjects, a similar tendency was discernible. For the lifting phase, no other effects of time or technique were revealed by the analysis of variance. In the lowering phase, the phase lag between hip and lumbosacral junction was considerably stronger when stoop lifting than when squat lifting (mean stoop lifting, 0.39 radians; mean squat lifting, 0.18 radians, $F(1, 49) = 39.0$, $p < .001$. In conclusion, the fatigue-related change in the coordination of lifting and lowering was limited to a decrease of the phase lag between knee and hip extension during the squat lift.

A second question in the present study was whether a systematic increase of the load on the low back occurs in repetitive lifting. To answer this question, first, we determined the maximum lumbosacral torques in both the lifting and the lowering phases of the movement. Table 3 contains the results for both lifting techniques. As can be seen, the average maxima did not increase in time. In addition, none of the subjects individually showed any evidence of an increase of the net lumbosacral torque. The peak torques during lowering were significantly higher in the stoop technique than in the squat technique. Also, a significant effect of trial on the peak torque during lowering was found ($p < .01$). The maximum torque averaged across subjects did not show a systematic trend in time, however. Nor did any of the individual subjects display such a trend.

A reduction of trunk extensor muscle performance was expected to cause an increased trunk flexion. The minimum lumbosacral angles measured during the successive lifting bouts are presented in Table 4. A low angle means a high degree of flexion. No effects of trial were found in either the lifting or the lowering phase of both techniques. The differ-

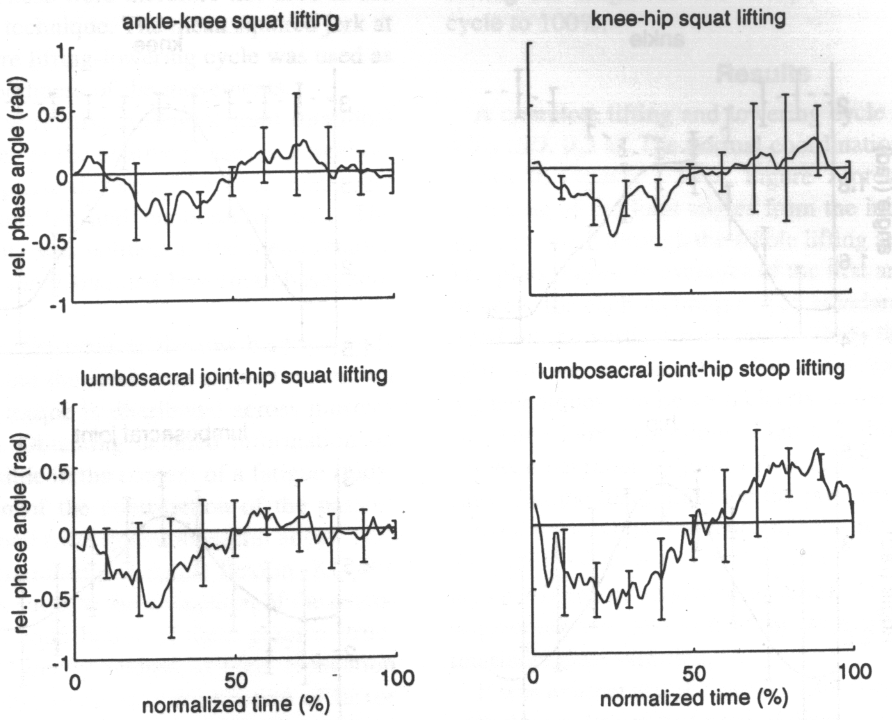

FIGURE 2. Mean relative phase of neighboring joints of 5 subjects. Top left, ankle–knee; top right, knee–hip joint; bottom left, hip–lumbosacral joint; and bottom right, hip–lumbosacral joint in stoop lifting. For stoop lifting, relative phases of the leg joints were not calculated, in view of the limited angular excursions. The bars indicate plus and minus one standard deviation.

ences between techniques were highly significant ($p < .0001$); more flexion occurred during stoop lifting.

Discussion

The only significant fatigue-related change in the coordination of lifting found in the present study was a decrease in the phase lag between knee and hip extension during squat lifting. Figure 7 shows this change in a plot of normalized knee angle versus normalized hip angle in the first and last analyzed trial. The relative retardation of the knee rotation with respect to the hip rotation could have been caused by fatigue of any of the muscles contributing to rotation of the lower leg. In view of the level of activity during lifting (Looze et al., 1993), fatigue of the gastrocnemius muscle might especially have played a role. A distoproximal sequence of joint rotations during lifting has been demonstrated previously (Burgess-Limerick, Abernethy, Neal, & Kippers,1995; Scholz, 1993b, Scholz, Milford, & McMillan, 1995). It has been hypothesized that the function of this sequence is to keep the biarticular muscles near optimum length for as long as possible (Burgess-Limerick et al., 1995). Because the distoproximal sequence of the rotation of hip and knee joints appeared to decrease systematically, this would imply a shift away from optimum muscle length and thus a reduced efficiency of the movement. In addition, optimum muscle length has been shown to shift to higher muscle lengths as a result of sustained activity (Huijing & Baan, 1995). This effect may cause the biarticular muscles to become active in a length range even farther from optimum length as a result of the decrease in phase lag alone.

The most striking finding in the present study was probably that changes in the performance of the lifting act throughout the 1/2-hr exercise were very limited. This lack of changes might be explained by a lack of fatigue induced by the protocol. However, several authors (Jørgensen, Andersen, Horst, Jensen, & Nielsen, 1985; Petrofsky & Lind, 1978; Potvin & Norman, 1993) have shown that lifting with similar loads and durations is certainly fatiguing for the low back muscles. The subjective reports of our subjects confirmed this; besides, they reported considerable fatigue in the lower extremities. Two subjects actually had to stop the exercise after the fifth instead of the sixth lifting bout because of excessive fatigue, 1 when stoop lifting and 1 when squat lifting.

Of course, many details in the execution of the lifting act will be missed when a two-dimensional linked segment analysis is used. First, Parnianpour and coworkers (1988) have shown in repetitive isoinertial trunk extensions that fatigue causes changes of the movement pattern mainly outside of the sagittal plane, that is, an increase of coupled movements mainly in the frontal plane. Therefore, we are making preparations to use a three-dimensional model in a

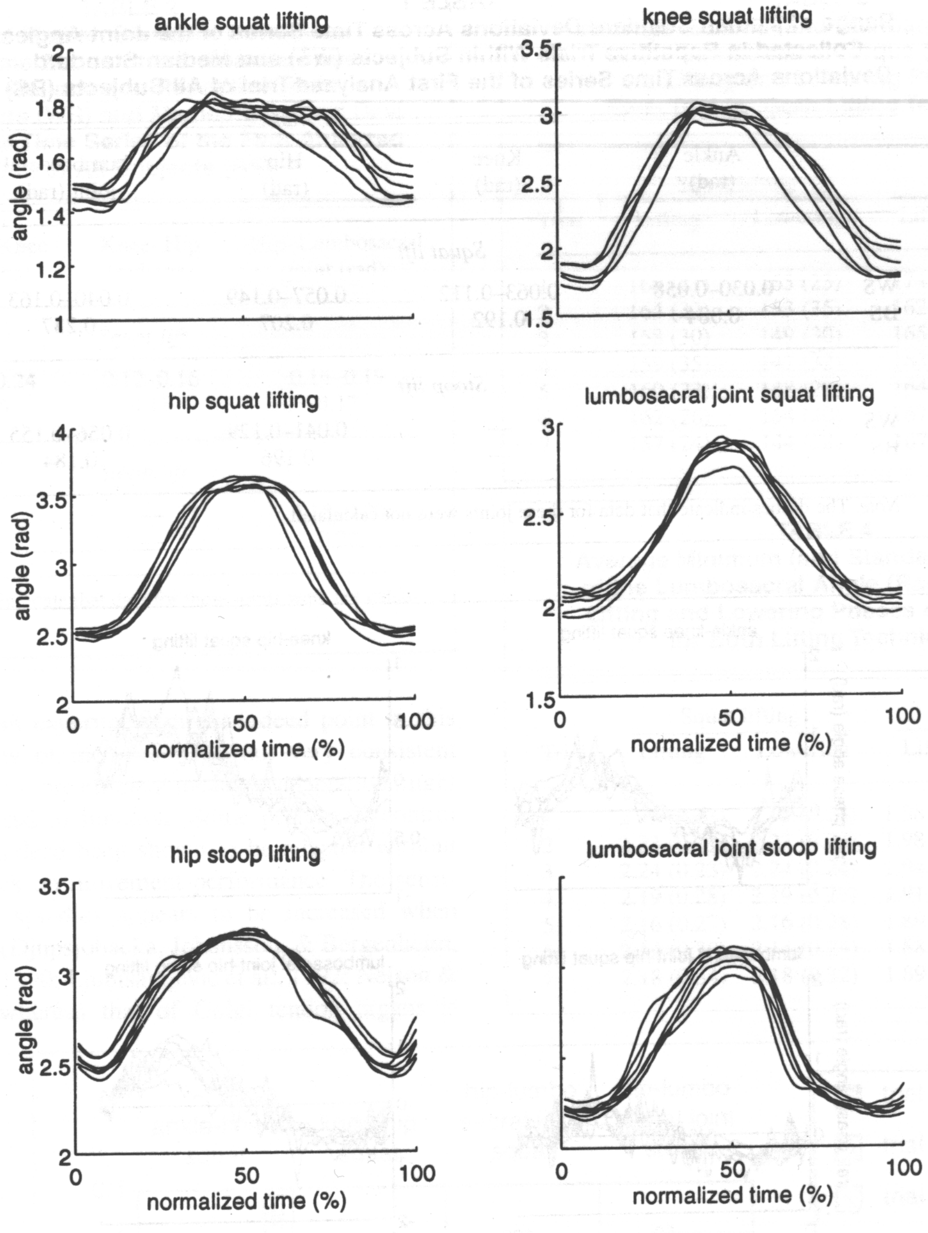

FIGURE 3. Upper four panels. Joint angle time series of repeated trials of a typical subject in squat lifting. Top left, ankle joint; top right, knee joint; middle left, hip joint; and middle right, lumbosacral joint. Lower two panels. Joint angle time series of repeated trials of a typical subject in stoop lifting. Left, hip joint; and right, lumbosacral joint.

similar study. Second, changes in the movement pattern at the level of a single or a number of motion segments of the spine will be missed by any method that treats the trunk essentially as a stiff segment. Third, changes in the pattern of the distribution of joint torques across muscles might occur. These could affect the load on passive structures but were not addressed in the present study. For instance, an increased coactivation of abdominal muscles during lifting would increase compressive loads on the spine. The latter effect seems quite likely to occur. Various studies have shown that fatigue-related types III and IV afference from muscles may through the γ system decrease the threshold of α-motoneurons of both the fatigued muscle and surrounding muscles, including antagonists (e.g., Djupsjöbacka, Johansson, Bergenheim, & Sjölander, 1995; Johansson, Sjölander, Sojka, & Wadell, 1987; Ljubisavljevic, Jovanovic, & Anastasiijevic, 1992). Preliminary evidence does indeed show that fatigued muscles tend to have increased cocontraction of antagonists (Gagnon, Arsenault, Smyth, & Kemp, 1992).

TABLE 1
Range of Median Standard Deviations Across Time Series of the Joint Angles Collected in Repetitive Trials Within Subjects (WS) and Median Standard Deviations Across Time Series of the First Analyzed Trial of All Subjects (BS)

	Ankle (rad)	Knee (rad)	Hip (rad)	Lumbosacral joint (rad)
		Squat lift		
WS	0.030–0.058	0.063–0.112	0.057–0.149	0.040–0.163
BS	0.084	0.192	0.207	0.237
		Stoop lift		
WS	—	—	0.041–0.129	0.056–0.155
BS	—	—	0.196	0.184

Note. The dashes indicate that data for these joints were not calculated.

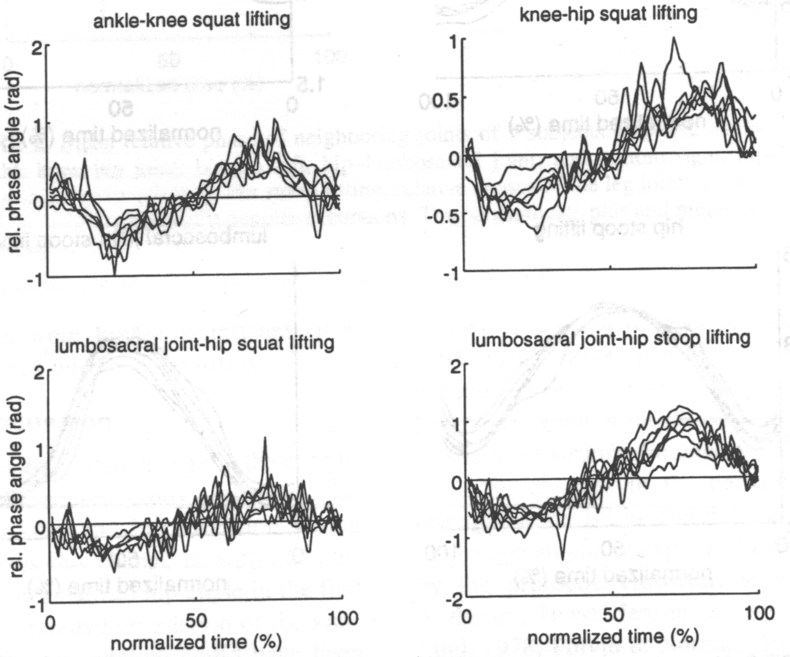

FIGURE 4. Relative phase-angle time series of repeated trials of a typical subject in squat lifting. Top left, ankle–knee; top right. knee–hip joint; bottom left, hip–lumbosacral joint; and bottom right, hip–lumbosacral joint in stoop lifting. For stoop lifting, relative phases of the leg joints were not calculated, in view of the limited angular excursions.

Even in view of changes that may have taken place but were missed by the present methodology, it seems quite remarkable that subjects were able, in spite of considerable fatigue, to maintain the desired movement pattern to a large extent. In line with this, only slight differences in the kinematics of repeated submaximal hopping have been shown to occur until the hopping height could no longer be maintained (Bonnard et al., 1994). It seems that the maintenance of such a stereotyped pattern in spite of strong changes of the input–output characteristics of the muscles places high demands on the adaptability of the neural control. Several authors (Arendt-Nielsen & Sinkjær, 1991; Bonnard et al., 1994; Lucidi & Lehman,1991) have shown that, in the presence of fatigue, timing of muscle activation is adapted to maintain torque, acceleration, and position profiles. If feedback were to control these adaptations on the basis of movement information, some disturbance of movement execution would be expected. The tendency toward an increase in

TABLE 2
Range of Median Standard Deviations Across Time Series of the Relative Phase Angles Collected in Repetitive Trials Within Subjects (WS) and Median Standard Deviations Across Time Series of the First Analyzed Trial of All Subjects (BS)

	Ankle–Knee (radians)	Knee–Hip (radians)	Hip–Lumbosacral joint (rad)
		Squat lift	
WS	0.13–0.24	0.12–0.16	0.14–0.19
BS	0.25	0.18	0.17
		Stoop lift	
WS	—	—	0.15–0.20
BS	—	—	0.25

Note. The dashes indicate that data for these joints were not calculated.

TABLE 3
Average Maxima (and Standard Deviation) of the Lumbosacral Torque (in Nm) in the Lifting and Lowering Phases of Each Trial for Both Lifting Techniques

	Squat lifting		Stoop lifting	
Trial	Lifting	Lowering	Lifting	Lowering
1	168 (21)	165 (23)	174 (37)	170 (35)
2	164 (32)	153 (35)	162 (19)	159 (30)
3	158 (30)	149 (30)	165 (30)	155 (28)
4	159 (35)	147 (32)	163 (28)	154 (33)
5	160 (23)	148 (29)	164 (30)	157 (29)
6	162 (26)	154 (30)	167 (33)	160 (39)
7	157 (24)	144 (28)	167 (35)	158 (38)

TABLE 4
Average Minimum (and Standard Deviation) of the Lumbosacral Angle (Radians) in the Lifting and Lowering Phases of Each Trial for Both Lifting Techniques

	Squat lifting		Stoop lifting	
Trial	Lifting	Lowering	Lifting	Lowering
1	2.18 (0.28)	2.13 (0.24)	1.98 (0.18)	1.96 (0.17)
2	2.23 (0.29)	2.23 (0.28)	1.98 (0.26)	1.98 (0.28)
3	2.24 (0.23)	2.24 (0.24)	1.94 (0.25)	1.94 (0.25)
4	2.19 (0.28)	2.19 (0.27)	1.91 (0.20)	1.91 (0.23)
5	2.16 (0.27)	2.16 (0.28)	1.89 (0.20)	1.90 (0.19)
6	2.14 (0.30)	2.14 (0.29)	1.88 (0.19)	1.88 (0.20)
7	2.18 (0.34)	2.18 (0.32)	1.89 (0.19)	1.88 (0.20)

jerk in the present experiment might indeed point in this direction. In view of the small and not very consistent changes, however, more efficient means of adaptation might have been expected to function. Some peripheral control properties have indeed been shown to be adapted without requiring feedback on movement performance. The sensitivity of muscle spindles appears to be increased when fatigue develops (Djupsjöbacka, Johansson, & Bergenheim, 1994; Gantchev, 1990; Ljubisavljevic et al., 1992; Nelson & Hutton, 1985), whereas that of Golgi tendon organs is

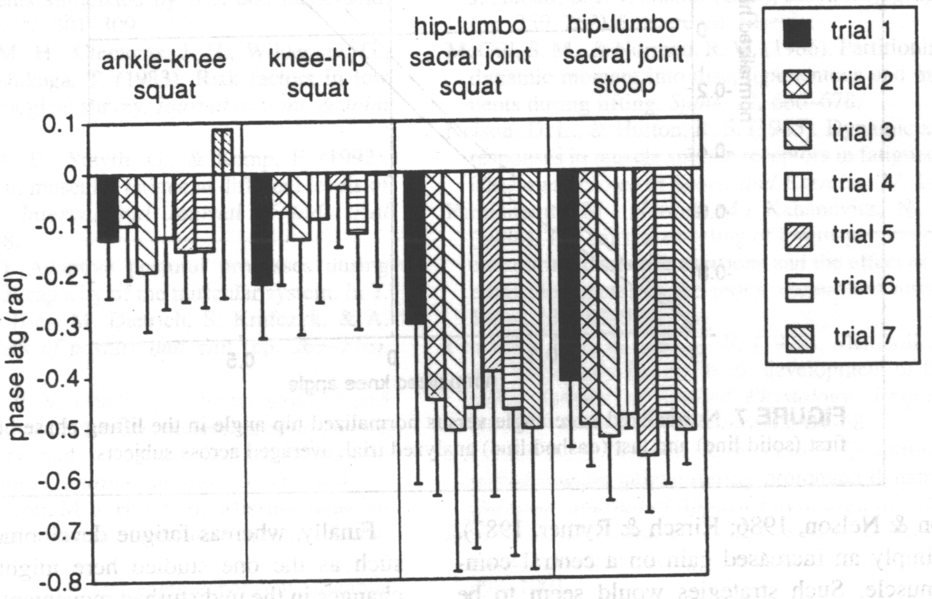

FIGURE 5. Mean phase lags between neighboring joints in lifting. The bars indicate one standard deviation.

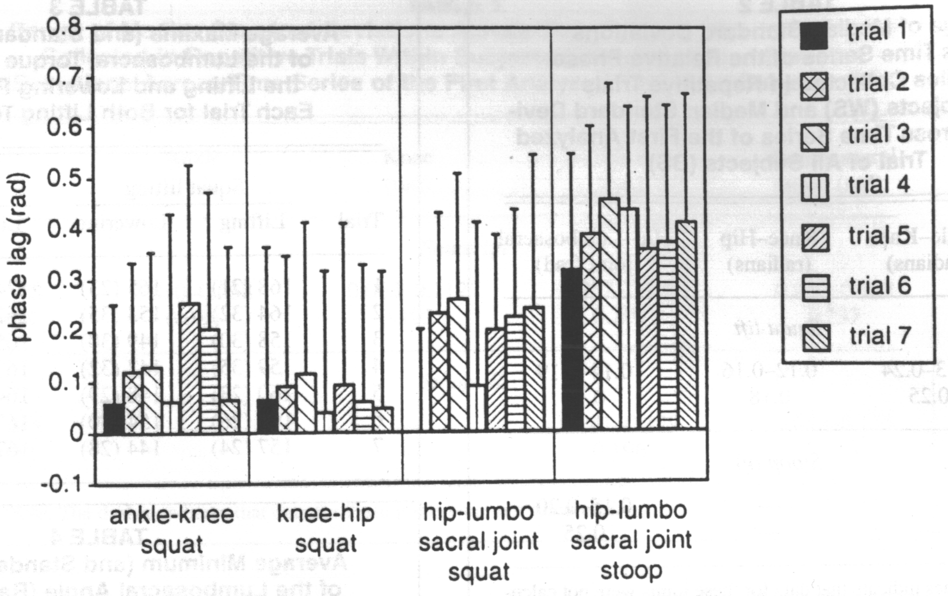

FIGURE 6. Mean phase lags between neighboring joints in lowering. The bars indicate one standard deviation.

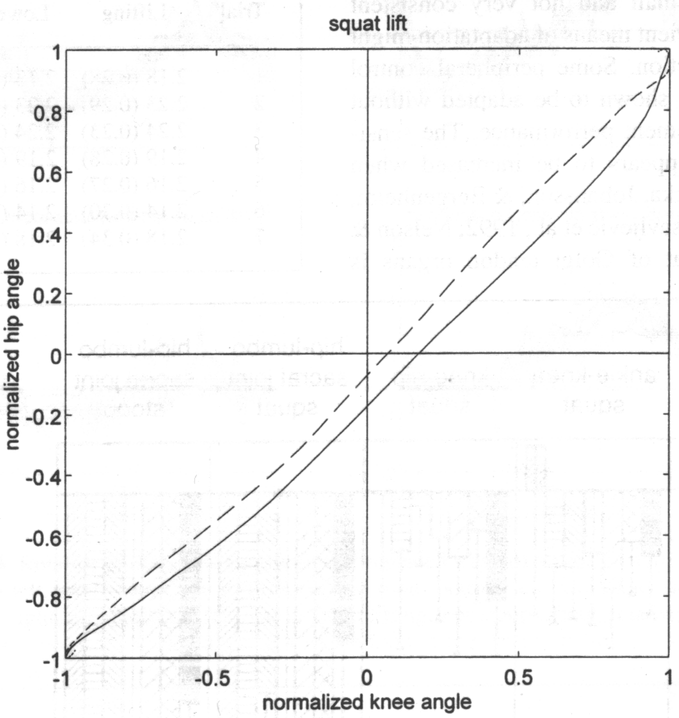

FIGURE 7. Normalized knee angle versus normalized hip angle in the lifting phase of the first (solid line) and last (dashed line) analyzed trial, averaged across subjects.

reduced (Hutton & Nelson, 1986; Kirsch & Rymer, 1987). Both changes imply an increased gain on a central command to the muscle. Such strategies would seem to be remarkably effective in maintaining the muscle output at the desired level without requiring that the central commands controlling the act be adapted.

Finally, whereas fatigue development in a repeated task such as the one studied here might lead to only minor changes in the undisturbed movement, it seems likely that it greatly reduces the possibility of reacting to perturbations. An increase of reflex motor time (Häkkinen & Komi, 1983; Kroll, 1974) has been reported to accompany muscle fatigue. Also, the rise time of voluntary trunk extension force has been shown to increase with fatigue (Parnianpour et al., 1988). These changes reduce one's ability to respond adequately and in a timely way to external perturbations, thereby possibly leading to excessive loads on structures in the low back. In addition, more prominent changes may occur when the lifting movement per se poses more maximal demands on the motor control strategy, for instance, when lifting at a high velocity or with a high load mass.

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